MEDICAL IMAGING GENESIS FOR FINITE ELEMENT-BASED MANDIBULAR SURGICAL PLANNING IN THE PEDIATRIC SUBJECT

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ABSTRACT- A CT imaging system has been used to develop a finite element (FE) model of the child-specific mandible. Correspondingly, a procedure to generate FE models from digital image data has also been developed. The geometric models are extracted from the CT scan data of children using ANALYZE AVW version 3.0, and then reviewed, edited and meshed using the preprocessor of DEFORM 3D version 6.0. By utilizing simplified material model and boundary conditions, detailed convergence tests were carried out using ABAQUS/STANDARD version 6.1. Test results show that over 52,592 linear tetrahedral elements or 37,062 degrees of freedom (DOF) are needed to model a child-specific mandible with reasonable accuracy. Results of the test series indicate that the surgical planning system is appropriate for further clinical implementation.

***Key Words*- Medical Imaging, Finite Element Model, Mandible, Craniofacial Deformity, Biomechanics**

I. INTRODUCTION

Numerical simulation can provide clinically useful information to the surgeon who must reconstruct pediatric craniofacial deformity. Such simulations should be done with clinically acceptable accuracy and in a reasonably short period of time. It is essential that finite element (FE) models be established from the non-destructive method based CT and/or MRI data of the patient [1, 2]. Current computer assisted craniofacial surgery planning systems are limited to anatomic restoration without regard to function [3]. Incorporation of finite element (FE) analysis as an integral component to simulate the stomatognathic function can complement and refine computer-assisted surgical planning.

While extensive studies on FE modeling of the human mandible have been carried out [1, 2, 4-7], these studies have largely focused on structural analysis of facial fractures in adults. Reconstructing a pediatric craniofacial deformity is fundamentally different. It involves repositioning the various elements of the craniofacial skeleton from an abnormal spatial relationship to restore the normal anatomic boundaries and function. In addition, the skeletal geometry and material properties in children differ from those of adults. This further warrants development of patient-specific FE models.

A critical step in the FE modeling process is the generation of the geometrical model of bony components. Compared to mechanical or fabricated parts, the geometric shape of the biomedical components is frequently complex and non-linear, especially the craniofacial complex, including mandible and maxilla. In such cases, it is difficult to describe the surface contour by a simple linear analytical model. In particular, the topology with multiple inner cavities and closed surfaces remains beyond the current capability of most commercially available CAD and FE codes. It remains a challenge for FE models to offer reasonable accuracy within a short period of time.

Since the accuracy of the finite element method can only be objectively established with a convergence test [1], the element size and the degree of freedom (DOF) must be determined on the basis of the convergence test results.

The objective of this study is to establish the methodology for routine development of child-specific mandibular models directly from medical imaging data. A convergence test is performed to specify mesh refinement needs and to estimate error.

II. METHODOLOGY

CT data was obtained from a helical scan using a GE Advanced CT scanner with a pixel matrix size of 512×512. The slice dimension (1.5 mm thickness) is determined by the standard of clinical practice. Using ANALYZE AVW 3.0 (Biomedical Imaging Resource, Mayo Foundation, Rochester, Minnesota), the data were reformatted and the voxels resized using trilinear interpolation, from 0.430×0.430×1.5 mm to 0.75×0.75×0.75 mm; the unassigned 16-bit file was converted to an unassigned 8-bit file. This procedure reduced the memory required for each file, from more than 200MB to 20 ~ 25 MB. Such data reduction does not markedly influence the quality of the 3-D images produced according to Lo *et al.* [3]. Under the volume render mode, hard tissue was separated from soft tissue by choosing a proper threshold value based on the tissue density. Skull and mandible were separated using the object separation technique. The adapt/deform method, one of the tile techniques, was used to generate the surface model.

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The .STL files of the mandible thus generated were then input into the preprocessor of DEFORM 3D (Scientific Forming Technologies Corporation, Columbus, Ohio). After editing any geometrical errors, a series of meshes with various numbers of nodes and elements were generated, primarily using the tetrahedral element with a nominal aspect ratio of 2.

The procedure for generating the geometrical model is shown in Figure 1. In this procedure, geometrical model editing, that is, the solid model repairing, cleaning and smoothing, is critical in evaluating the non-linear geometry of the mandible. Geometrical model repairing refers to the elimination of the geometrical discontinuities; while cleaning refers to the removal of inner cavities and closed surfaces. Otherwise, meshing may not be successful. Surface smoothing reduces the minor geometrical distortion during digitizing. In this study, geometrical model editing was carried out manually and surface smoothing was limited to obviously distorted points. Our geometrical model editing will likely benefit from continuing efforts to develop automated techniques.

The .KEY files generated by DEFORM 3D were rewritten as .INP files (ABAQUS/STANDARD code). Boundary conditions consisted of a simplified muscle force matrix, as shown in Figure 2. All degrees of freedom (DOF) for the 6 nodes at each condyle process were constrained to simulate the temporomandibular joint (TMJ) reactions normal to the articular eminence and to prohibit rigid-body motion of the mandible [1, 5]. A load of 100 N was applied at the center of the incisors in a direction perpendicular to the occlusal plane.

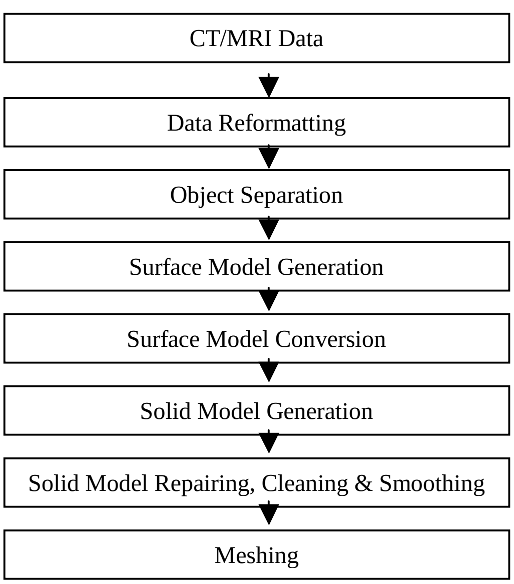

Figure 1. Procedure for generating geometrical models and meshing.

A simplified material model was postulated. Thus, the entire mandible was assumed to be homogeneous, isotropic, and linear elastic, with Young’s modulus of 13.7 kN/mm^2, or 13.7 GPa, and Poisson’s ratio of 0.3 [6].

In order to estimate the error in the solution when a 4-node linear tetrahedral element was employed, a series of FE analyses were carried out using ABAQUS/STANDARD 6.1 (Hibbitt, Karlsson & Sorensen Inc., Providence, Rhode Island). The only difference in the various analyses was the mesh density of the FE models. Some typical parameters including the maximum and minimum principle stresses, Von Mises stress, displacements, U_1, U_2 and U_3, referenced to the global coordinate system, and magnitude of the total displacement, U, were checked at selected points. The points selected to check stresses were at the posterior border of the ramus, the sigmoid arch and the mandibular body. Points selected to check the displacement were those at the angle, the coronoid process, the first molar and the bottom of the chin, as shown in Figure 2.

III. Results

The number of nodes and elements employed in each analysis along with a portion of the study results are listed in Table I. The variation of the maximum principle stress in the mandible is shown in Figure 2.

The Von Mises stress and maximum and minimum principle stresses at the checked points do not significantly change with an increase in the number of elements within the range employed for the current study.

On the other hand, the variation of displacements with the number of elements yields the typical response curve representing a bent ‘knee’ in the graphical output [8]. At that point, the number of elements is close to 52,592, as shown in Figure 3. The size of the elements corresponding to the element number of 52,592 was about 1.2 mm, while there were 37,062 DOF. This is close but somewhat larger than the 30,000 DOF concluded by Hart *et al.* [1]. It is a reasonable result as the 20-node hexahedral element was employed as the principal element in their study while the 4-node tetrahedral element was used in the current study.

IV. Discussion

The accuracy of the generated geometric model of the mandibular element of the craniofacial skeleton was evaluated while considering following points:

(1) The subjects in this study are actual patients rather than cadaveric dry specimens [4, 6] or wet specimens [1, 7] as used in some studies.

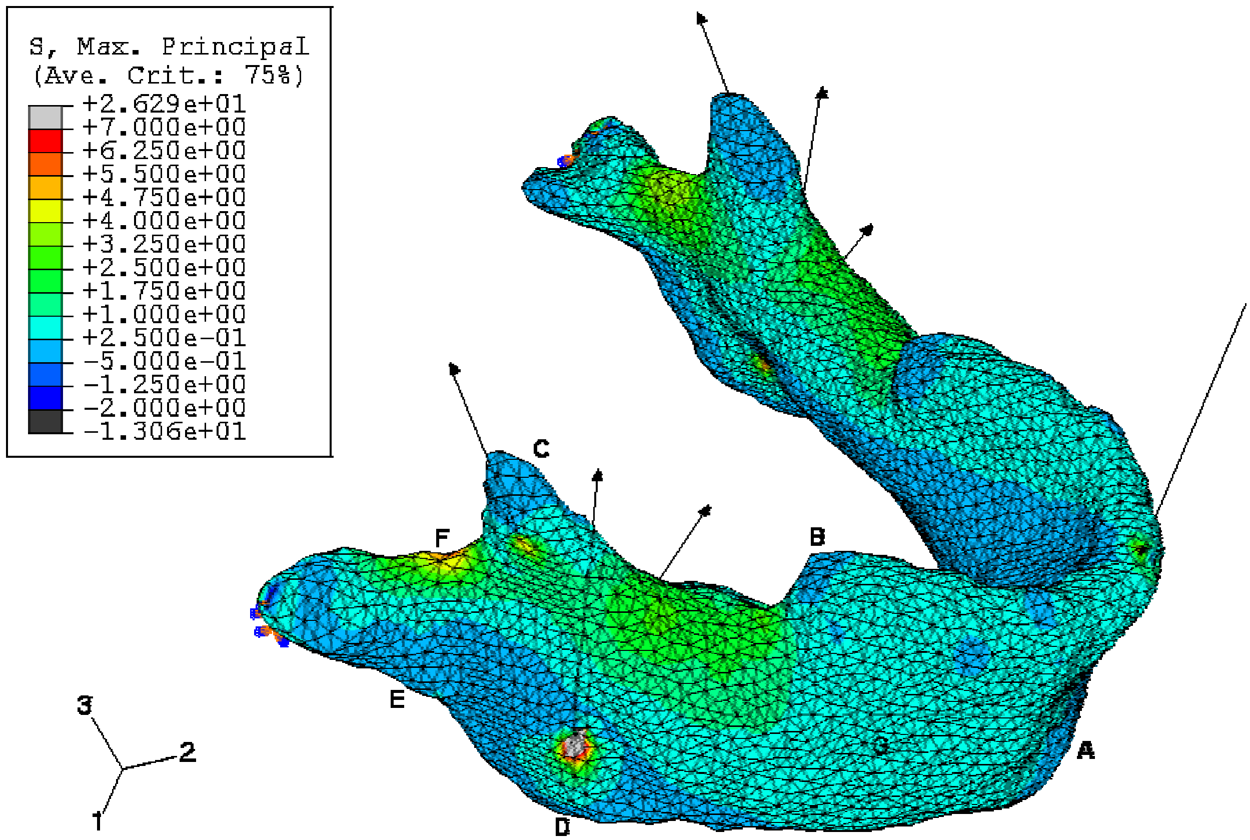

Figure 2. A plot of distribution of the maximum principle stress with 67,190 elements, boundary conditions and loading, and selected points.

TABLE I
CONDITIONS AND RESULTS OF THE CONVERGENCE TEST

Number of Nodes	Number of Elements	Number of Degrees of Freedom (DOF)	Von Mises Stress at Mandible Body	Maximum Principle Stress at Mandible Body
			MPa	MPa
1050	3562	4150	0.418	0.156
3303	12756	9909	0.470	0.239
5233	20926	15699	0.492	0.274
7194	29350	21582	0.487	0.264
9439	39470	28317	0.489	0.285
12354	52592	37062	0.510	0.297
15590	67190	46770	0.486	0.284
19064	82632	57192	0.507	0.295
21813	95209	65439	0.509	0.297

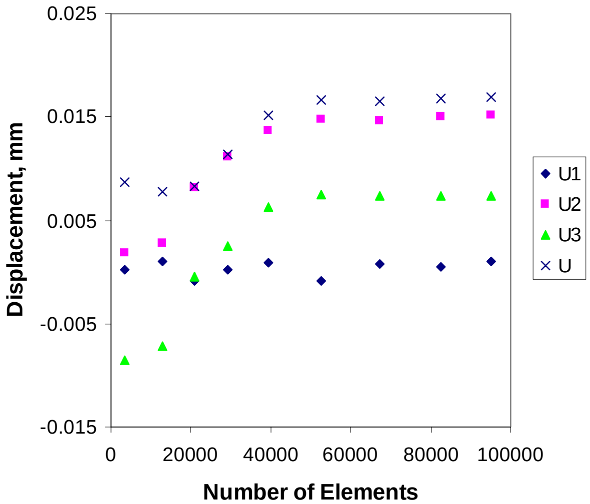

Figure 3. Global coordinate displacements at right first molar (point b) as a function of the number of linear tetrahedral elements.

(2) Generally, the smaller the slice thickness of the CT scan, the more accurate the geometric model. In this study, the slice thickness was 1.5 mm, which was less than 2 mm [4], 3 mm [1] and 10 mm [6] thickness reported in the literature. (Further reduction of the slice thickness of the CT scan to 1 mm is limited due to prolonged CT scan time introducing motion artifact and an increased risk to radiation exposure to the child.)

(3) The surface model was manually repaired and cleaned of extraneous errors by making use of the graphic interface followed by a smoothing of the topography. This allowed both operator evaluation and advanced graphic tools and representations. However, the overall process remains time and labor intensive. Approximately 72 hours is needed to model the mandible alone. It is expected the ongoing code development will reduce future working time. The objective of our imaging laboratory is to develop a clinically applicable FE model of the craniofacial skeleton relevant to the practicing surgeon within an acceptable pre-surgical planning time frame.

V. Conclusion

A procedure to rapidly generate patient-specific FE models from CT scan data was developed and a FE model of the child-specific mandible is evaluated. Convergence testing indicated that over 52,592 four-node tetrahedral elements are needed to obtain an accurate child-specific mandibular model.

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